THE INFLUENCE OF ROCKER PROFILE FOOTWEAR ON ROLLOVER DURING WALKING

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THE INFLUENCE OF ROCKER PROFILE FOOTWEAR ON ROLLOVER DURING WALKING

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LIST OF SYMBOLS AND ABBREVIATIONS

AFO	Ankle Foot Orthosis combined with integrated footwear system
No AFO	Integrated footwear system but absence of Ankle Foot Orthosis
FREE	Ankle Foot Orthosis in free ankle plantarflexion and dorsiflexion with integrated footwear system
STOP	Ankle Foot Orthosis in maximal ankle constraint limiting plantarflexion and dorsiflexion with integrated footwear system
STOP-NR	Ankle Foot Orthosis in maximal ankle constraint but absence of integrated footwear system

SUMMARY

Rocker profiles are one of the most commonly prescribed footwear modifications provided to individuals with impaired rollover. Impaired rollover is caused due to loss of neuromuscular function (i.e. stoke) or orthotic ankle constraint. When rollover is impaired, continued forward progression is interrupted and walking gait becomes less efficient (i.e. increased energy expenditure). Rocker profile footwear modifications are designed to mimic the functions of the anatomical ankle-foot rockers and provide its users with a smooth and efficient rollover. However, while there is theory governing the design of a rocker profile and subjective descriptions of rocker profile function, the extent to which a rocker profile footwear provides rollover has not yet been quantified. The aim of this study was to quantify effective and ineffective rollover and test whether our rocker profile provides effective rollover. We hypothesized that healthy subjects (n=4) walking with orthotic ankle constraint and the rocker profile (STOP) would have no change in rollover and energy expenditure outputs compared to walking with orthotic ankle free and rocker profile (FREE); but that healthy subjects (n=4) walking in STOP would have a change in rollover and energy expenditure outputs compared to walking with orthotic ankle constraint and no rocker profile (STOP-NR). To test this hypothesis, rollover was quantified as stance phase duration, cadence and radius of curvature and energy expenditure was quantified as heart rate and rating of perceived exertion. In addition to these outputs, we analyzed the ground reaction forces and duration of stance in early, middle and late stance period to determine the effects of the rocker profile footwear components. Through the rollover and energy expenditure outputs of the STOP, FREE and STOP-NR conditions, we quantified effective rollover as 0.29 (0.01) radius of

curvature with a heart rate of 110.5(6.7) bpm and ineffective rollover as 0.69(0.12) radius of curvature with a heart rate of 131.5 (8.1) bpm. By creating this scale, we were able to determine that our rocker profile provided effective rollover (0.34[0.04] radius of curvature with a heart rate of 111.3[8.3] bpm). However, a future study with a greater sample size is needed to confirm these results.

CHAPTER 1

INTRODUCTION

Applications of rocker profile footwear

Footwear that contains a rocker profile is one of the most commonly prescribed therapeutic lower limb motion control technologies (Hutchins 2009). The "rocker profile" describes the rounded shape of the external sole of the footwear (Hutchins 2009). Fundamental characteristics of rocker profile footwear are the rounded shape of the sole which provides rolling motion. Clinicians often attempt to use rocker profile footwear as a treatment to restore rolling motion of the lower limb in persons with impairment of ankle function due to disability (i.e., restricted ankle motion due to osteoarthritis, limited ankle motion due to stiffness created by neuromuscular spasticity, impaired ankle motion due to coalesced tarsals in diabetic neuroarthropathy, etc.). In other cases, rocker profile footwear is used to restore rolling motion in persons that use ankle foot orthoses (AFOs) when the AFO is designed to constrain ankle motion (i.e., AFO that constrains ankle motion in a person with muscle weakness due to paraplegia). In these cases, constraint of ankle motion due to physical impairment or orthotic ankle constraint the function of the ankle-foot complex is compromised and normal ankle motion is restricted. The restriction of normal ankle motion interferes with gait by interrupting rollover dynamics during stance phase (Hovorka 2013, Farris 2012). In these instances, rocker profile footwear may be used to provide rolling motion during stance phase of gait to restore roll over dynamics. Roll over dynamics during stance phase of gait have been described as occurring through coordinated functions of ankle-foot rockers beginning at the heel and progressing to the ankle and toes through four rocker phases that provide lower limb roll over and forward advancement during gait (Hutchins 2009, Hansen 2010).

Understanding rollover and the four rockers of gait

Rocker profile footwear designs attempt to "mimic" the function of the intact ankle foot complex. The ankle-foot complex consists of a series of articulations between the bones within the shank and the bones within the foot. The bones within the shank are the tibia and fibula, and the bones within the foot consist of 7 tarsals, 5 metatarsals, and 14 phalanges. Thirty three articulations produce complex motions that are defined as the end points of the bones **(Figure 1)**. Motions at the articulation are produced by over 100 muscle-tendon structures in three cardinal planes: sagittal, coronal and transverse. Additional structures (skin, fascia, ligaments, fat pad, etc.) also contribute to motions at the articulations. In analyzing the performance of the foot and ankle complex during gait, biomechanists typically quantify and characterize the motion of these articulations in the sagittal plane as flexion and extension because these motions account for the majority of motion to produce stable and uniform gait (Perry 2010). Moreover, the coordinated function of the foot and ankle complex during stance phase of gait has been described as a series of four rockers that produce a "rolling" function. These four rockers are the heel rocker, ankle rocker, the forefoot and toe rocker (**Figure 2**). During the early stance phase of gait, the heel rocker serves as the first of four rockers and is engaged during two periods of the gait cycle: initial contact and loading response. During initial contact (0-2% gait cycle), the heel rocker provides "braking' or deceleration of lower limb motion as the heel of the foot impacts the ground (Perry 2010). As the foot progresses toward the ground and advances in stance phase, loading response occurs. During loading response (2-12% gait cycle), the heel rocker continues to decelerate the limb and advances the forefoot to achieve foot flat (Perry 2010). Once the length of the foot from heel to toe is in contact with the ground, loading response is concluded. After the foot is in contact with the ground, the fulcrum moves from the heel to the ankle, thus engaging the ankle rocker. As the second rocker, ankle rocker occurs during the middle period of stance (12-31% gait cycle). During the mid-stance period, ankle rolling motion produces dorsiflexion which allows forward progression of the shank and body weight acceptance (Perry 2010). As the shank progresses forward and the ankle continues rolling into greater dorsiflexion, the calf muscle tendon unit (medial and lateral gastrocnemii, soleus and Achilles tendon) undergo tension. Increasing tension in the calf muscle tendon unit halts ankle dorsiflexion and causes the heel to rise, which moves the fulcrum from the ankle to the metatarso-phalangeal joints of the forefoot. The initiation of heel rise marks the end of ankle rocker and the beginning of forefoot rocker. During the forefoot rocker (also known as third rocker) which occurs during late stance (31-50% gait cycle), is characterized by a peak in the ground reaction forces (Perry 2010). The metarso-phalangeal joints of the forefoot serves as a fulcrum which allow transfers of ground reaction forces from the stability of double limb support to the initiation of "propulsion". Thus, forefoot rocker is initiated via tension in the calf muscle tendon unit that produces heel rise and the forward progression of the ground reaction force toward the toes. As the lower limb continues to advance forward, the fulcrum moves forward to the toes to engage the interphalangeal joints which marks the initiation of toe rocker when the leg is in and single limb support. As the fourth and final rocker, toe rocker occurs during pre-swing of the gait cycle (50-62% gait cycle) and serves as the platform for which lower limb "propulsion" occurs. During fourth rocker, the lower limb plantarflexor and dorsiflexor muscles are activated and the lower limb is lifted off the ground, thus terminating stance phase and initiating swing phase.

In healthy subjects, the ankle-foot rockers produce controlled progression of the lower limb during stance phase. Effective rollover of the lower limb has been qualitatively characterized as a smooth forward progression of the shank and low energy expenditure (Adamczyk 2006, Adamczyk 2013, Hansen 2008). When movement of the ankle-foot complex is impaired due to mechanical constraint (lower leg fracture cast, ankle foot orthosis) or physical impairment (muscle paralysis due to stroke, advanced diabetes, etc.), the smooth forward progression of the lower limb is interrupted thus compromising rollover. Compromised rollover may interrupt forward progression, produce gait compensations and increase energy expenditure (Vanderpool 2008, Waters 1982, Mattson 1990, Fowler 1993).

Figure 1. Osseous anatomy of the lower limb.
The osseous anatomy of the ankle-foot complex represents the terminus of the lower limb. It consists of two bones in the shank that articulate with over 30 bones in the ankle-foot complex.
Web site URL: https://www.opensesame.com/c/foot-and-ankle-anatomy-healthcare-professionals-training-course Accessed on May 4, 2014

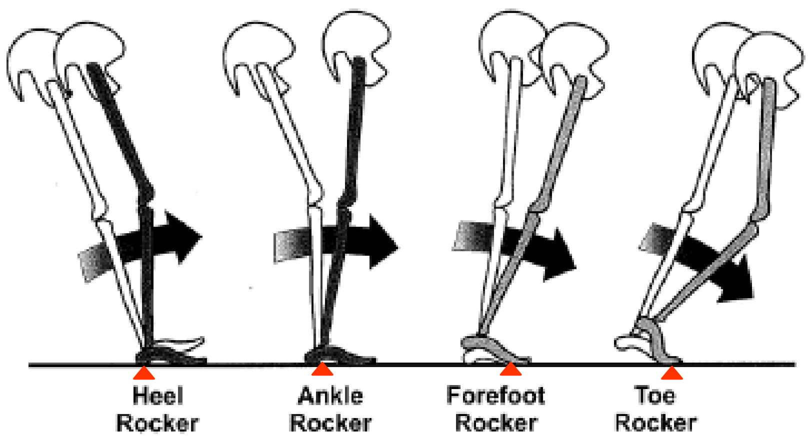


Figure 2. The four rockers of the ankle-foot complex.
Healthy ankle-foot complex engages as a series of rockers to provide rollover. Red triangle indicates location of fulcrum at the heel in heel rocker, at the ankle in ankle rocker, at the metatarso-phalangeal joints in forefoot rocker and at the toes in toe rocker. Figure modified from Perry J, 2010

Quantifying rollover during gait

The characteristic features of rollover have been subjectively described in the literature, but have not been clearly quantified (Hutchins 2009, Hullin 1991). The most notable quantification of rollover function and the performance of the lower limb rockers has been examined as an inverted pendulum model of walking. The inverted pendulum model describes the function of the ankle-foot complex as a “rolling wheel” during stance phase (McGeer 1990). Based on the inverted pendulum theory of gait, biped robots with curved ankle-foot segments and point center of masses were used as simple models to study the dynamics of walking (McGeer 1990, Kuo 2005). Through these studies, it was determined that the optimal curvature for the biped robots to achieve rollover for steady state walking was a radius that was 30% of the lower limb length from the greater trochanter of the hip to the bottom of the foot. This finding has been examined and confirmed to be the same radius of curvature in human walking gait analysis studies (i.e., humans have a radius of curvature that is 30% of their limb length) (Hansen 2000). Recent studies by Hansen, determined that the ideal “rollover shape” for human walking in a variety of incline and speed conditions was 30% of limb length (Hansen 2008, Hansen 2012). Similar studies by Hansen also examined the “rollover shape” of subjects as a controlled variable and it was determined that given various shoe radii, humans altered their joint angles to achieve the desired radius of 30% limb length (or 0.3) (Hansen 2010). However, due to the minimal stability that a curved rocker profile footwear system provides and the difficulties in achieving the transition from standing to walking gait, rocker profile in footwear used by persons with movement disorders (i.e., persons with physical impairments that render unstable gait such as stroke, spinal cord injury, lower limb fracture, etc.) are not completely curved to 30% limb length (Hutchins 2009, Hillin 1991). Instead, typical rocker profiles are designed to provide more stable geometries (i.e., radius of curvature greater than 30% of limb length) and the footwear is made with deformable materials that compress under loading to produce radius of curvature greater than 30% of limb length (Arazpour 2013).

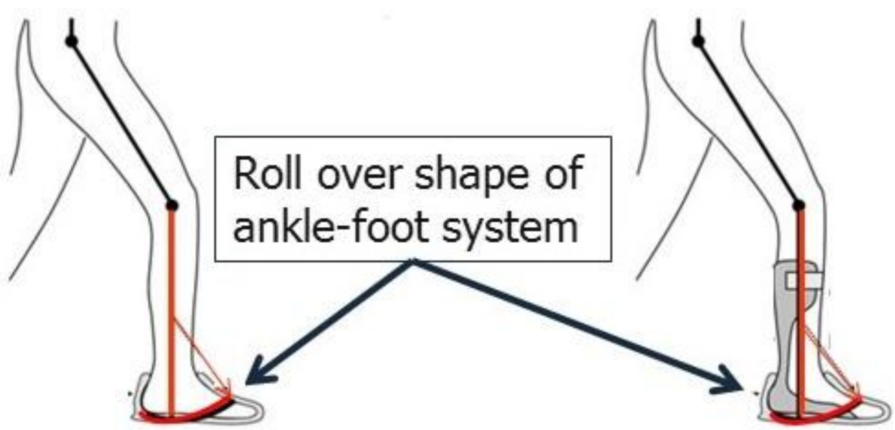


Figure 3. Roll over shape of the lower limb during stance phase.
The ankle foot complex produces a roll over shape during stance phase that traverses an arc with radius of curvature. **(a)** Roll over radius of curvature calculated when subjects walk without constraint of lower limb motion and **(b)** roll over radius in orthotic ankle constraint in an AFO-footwear combination. Adapted from Hansen A, Lower Extremity Review, 2010

There are a variety of rocker profile footwear designs (i.e., heel to toe rocker, double rocker, negative heel, etc.) that provide roll over. The efficacy of these rocker profiles footwear designs to enhance or restore lower limb forward progression have been examined in human subject gait analysis studies in which lower limb movement trends were quantified in response to various rocker profile designs (Hutchins 2009). These studies have quantified differences in subjects gait as limb motion (joint angles), forces (joint torques), temporal spatial (stance and swing phase duration, step length) and roll over radius (radius of curvature due to walking with a variety of rocker profile footwear designs (Hutchins 2009). However, these studies were not performed in the context of impaired ankle-foot motion and as a result they did not examine the effect of rocker profile footwear in persons that may not be able to produce adequate gait compensations. Also, the results from these studies were inconclusive and offered no definite characterization of the effects of rocker profile footwear on a person's rollover performance.

Background

Based on the review of research literature regarding optimal rocker shape and previous rocker profile footwear designs, we designed and developed a custom rocker profile footwear system for use in an experimental paradigm which aimed to preserve the functions of the ankle-foot rockers when the footwear is used in combination with of an ankle foot orthosis (AFO) designed to control ankle motion. Previous studies conducted in our lab were aimed at analyzing the effectiveness of rocker profile footwear in maintaining rollover. The results of those studies revealed that the rocker profile footwear used in an AFO that maximally constrained ankle motion was effective in maintaining stance phase duration, step length, stride length and radius of curvature to the same levels achieved when walking with no AFO (Oludare 2012). The stance duration of subjects walking in orthotic ankle constraint with rocker profile was approximately 60%, the same stance duration as in the no orthotic ankle constraint condition. Step length during the orthotic ankle constraint condition walking with rocker profile was approximately 0.8m, which was the same as during unconstrained walking. The radius of curvature in subjects walking in the orthotic constrained condition with rocker profile compared to no orthotic constraint was approximately 0.28 and 0.25 respectively, a difference of about 11%. These studies also revealed that there was no knee and hip joint compensation due to the lack of ankle-foot motion (Oludare 2013). There was less than 5% degrees of peak extension and flexion for the knee joint and hip joint in the constrained walking with rocker profile compared to the unconstrained walking condition. However, these studies led to additional questions about defining the rocker profile performance which motivated this study. The primary questions which arose were: 1) whether the rocker profile was primarily responsible for preserving rollover while walking without the ankle-foot complex, and 2) which components of the rocker profile footwear were most important in preserving rollover while walking with a constrained ankle-foot complex. Based on these questions, we have designed a novel experimental paradigm which examines rollover in a condition which eliminates the ankle-foot complex without the rocker profile, and we have also developed a method for analyzing the function of each of the rocker profile footwear's components. From the results of this study, we will quantify the range of rollover and create a system which clinicians and rocker footwear designers will use to characterize the performance of footwear designs.

Hypotheses

<u>Hypothesis 1</u>: We hypothesize that rollover and energy expenditure of healthy subjects walking in rocker profile footwear and no orthotic ankle constraint (No AFO) will not change when walking in orthotic ankle free and rocker profile footwear (FREE).

Hypothesis 2: We hypothesize that rollover and energy expenditure of healthy subjects walking in AFO free will change when walking in orthotic ankle constraint and footwear without rocker profile (STOP-NR).

Hypothesis 3: We hypothesize that rollover and energy expenditure of healthy subjects walking in rocker profile footwear and no orthotic ankle constraint (STOP) will not change in FREE, but rollover and energy expenditure in AFO stop will change when walking in STOP-NR.

To evaluate rollover (stance phase duration, cadence and radius of curvature) and energy expenditure (heart rate and rating of perceived exertion), subjects walked in a motion analysis laboratory in the following conditions: footwear but with no AFO, footwear combined with FREE which allowed free ankle plantarflexion and dorsiflexion, footwear combined with STOP which maximally constrained ankle plantarflexion and dorsiflexion and no rocker profile footwear combined with AFO in maximal constraint of ankle plantarflexion and dorsiflexion (STOP-NR).

To quantify rollover, we calculated the stance phase duration, cadence and radius of curvature of each subject using ground reaction forces and a marker model. To quantify energy expenditure, we collected each subject's heart rate and rating of perceived exertion.

CHAPTER 2

METHODS

Subjects

Subjects participated in a dynamic loading study to quantify rocker profile footwear function. Four healthy individuals [3 male, 1 female, mean (standard deviation) age 20.4(1.34) years, height 175.0(8.12) m, mass 70.9(8.49) kg, preferred walking speed 1.41(0.08) $m \cdot s^{-1}$] gave written informed consent prior to their participation in a protocol approved for human subject research by the Institutional Review Board of the Georgia Institute of Technology. Subjects reported no orthopedic, neuromuscular or gait abnormalities.

High performance AFO

A custom high performance motion limiter AFO was created to achieve desired control of ankle motion (Hovorka 2014, in progress). The high performance AFO contained rigid carbon composite shank and foot sections molded into full cylindrical shell. The diameter of shank and foot cylindrical shells was of sufficient size to fit a range of limb dimensions. The anterior shank cylinder contained an articulated tibial plate with threaded screw attached to a ball and socket joint to allow customization of fit to each subjects' shank. Straps containing contoured pads located at the dorsum of the foot and apex of the heel provided customization of fit to each subjects' foot. Anchors attached to the ends of a linear bearing ankle motion controller maximized moment arm length with their location very near the end of each shell. Modular clamps were attached onto the linear bearing to stop ankle plantarflexion and dorsiflexion to create a STOP condition that maintained shank to floor angle in mid-midstance (Owen 2004, Owen 2010). Mid-midstance alignment was determined with subjects wearing AFO in relaxed standing while on a loading platform (LASAR Posture, Otto Bock Health Care Inc., Minneapolis, MN) that projected the ground reaction force vector in the sagittal plane through the subjects' knee center and through a bisection of the foot (Blumentritt 1997, Owen 2004, Owen 2010). To create the FREE condition, clamps were removed from the linear bearing to provide free ankle plantarflexion and dorsiflexion motion.

The efficacy of the AFO in constraining the ankle-foot complex has been documented in previous studies and upcoming publications (Hovorka 2014, in progress). In a static loading experiment, the AFO in the orthotic ankle constraint condition reduces the ankle motion by 97% in plantarflexion and by 80% in dorsiflexion when compared to the AFO in orthotic ankle free condition (Hovorka 2014, in progress). During dynamic loading experiments where healthy subjects walked on a dual belt treadmill, the AFO in the orthotic ankle constraint condition reduced ankle motion by 79% in plantarflexion and 88.6% in dorsiflexion (Hovorka 2014, in progress). In the static and dynamic trials, subjects had insignificant differences in ankle-foot plantarflexion and dorsiflexion when the AFO in the orthotic ankle free condition was compared to the no AFO condition.

Rocker profile footwear

The rocker profile footwear created for this study was designed to provide high performance rollover. To achieve this, the rocker profile footwear was designed to mimic the ankle-foot function during early, middle and late stance phase in three sections: cushion heel, midfoot platform, and forefoot/toe rocker (**Figure 4**). The heel portion is composed of top, middle and bottom layer consisting of compliant materials based on the theory that it will compress after initial contact and descend the foot toward the ground (Marzano R 2002, Caron M 1999, Baker PL 1970, Cracchiolo A 1979, Zamosky I 1964). This would simulate first rocker by providing pseudo ankle plantarflexion during early stance phase (Hutchins 2009, Perry 2010). The firm midsole design is composed of a top, middle and bottom layer consisting of less compliant materials than the cushion heel. The less compliant midfoot platform should theoretically compress minimally under load to allow continued forward progression of the lower limb that simulates second rocker and provides pseudo ankle dorsiflexion during middle stance phase (Hutchins 2009). The forefoot design is composed of a top, middle and bottom later which include a middle layer fulcrum located near the metatarsal heads with a radius equivalent to the first metatarsophalangeal joint. In addition, a bottom layer flexible hinge and toe ramp with an 11° incline. The location of the forefoot fulcrum apex is at 60% of sole length from posterior heel respectively to provide for stance phase rollover. Thus, the forefoot/toe rocker section of the rocker profile should theoretically allow pseudo ankle plantarflexion and metatarsophalangeal motion (Van Bogart 2005) to engage third and fourth rocker during terminal stance (Hutchins 2009, Perry J 2010).

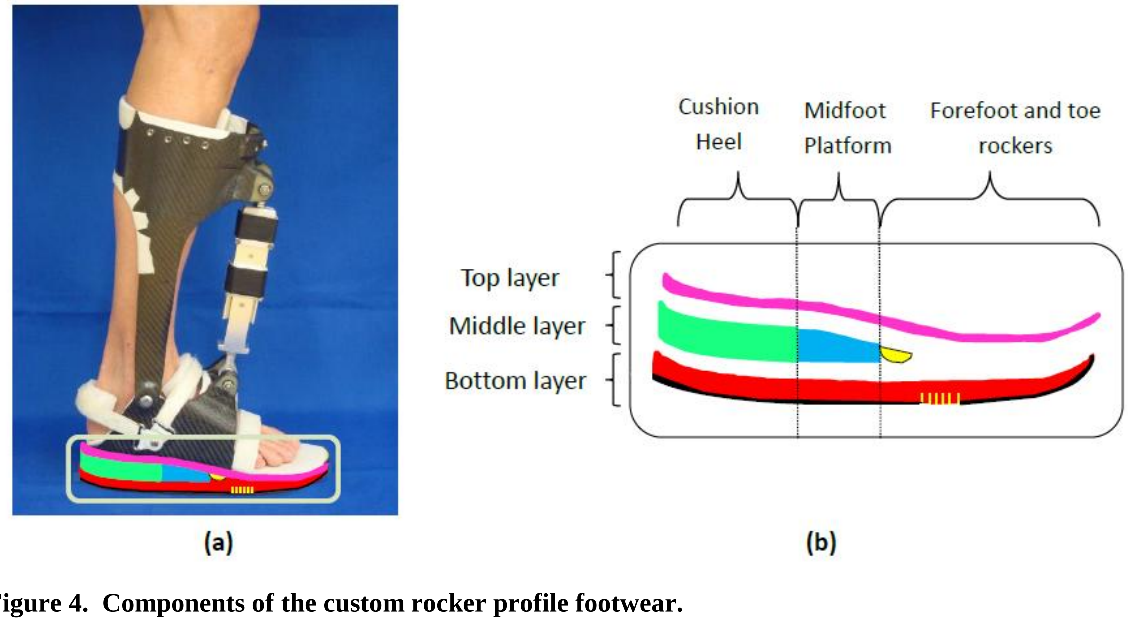


Figure 4. Components of the custom rocker profile footwear.
Rocker profile footwear is anchored to the bottom portion of the foot shell to create an integrated footwear system. Footwear sections represented as colored regions. **(b)** Components of integrated rocker profile footwear: top layer contains moderately rigid cork from heel to toe (purple); middle layer contains compliant foam at heel (green), rigid cork at midfoot (blue) and fulcrum (yellow); bottom layer contains compliant foam sole (red) from heel to toe with toe ramp and fenestrated metatarsal region (yellow vertical lines). Exterior rubbers tread (black).

Rigid footwear without rocker profile

Rigid footwear without rocker profile was created for this study to provide low performance rollover. To achieve this, the rigid footwear was designed with a flat sole which compressed minimally compared to the rocker profile footwear (**Figure 5**). The rigid footwear was designed to be of similar size (length, width and height), and mass as the rocker profile footwear. The rigid footwear consists of carbon fiber composite material that was fabricated around a positive model of the rocker profile to ensure that the sizes of the rigid footwear and the rocker profile were similar. After the lamination, the positive model was removed which created a hollow interior and rigid exterior.

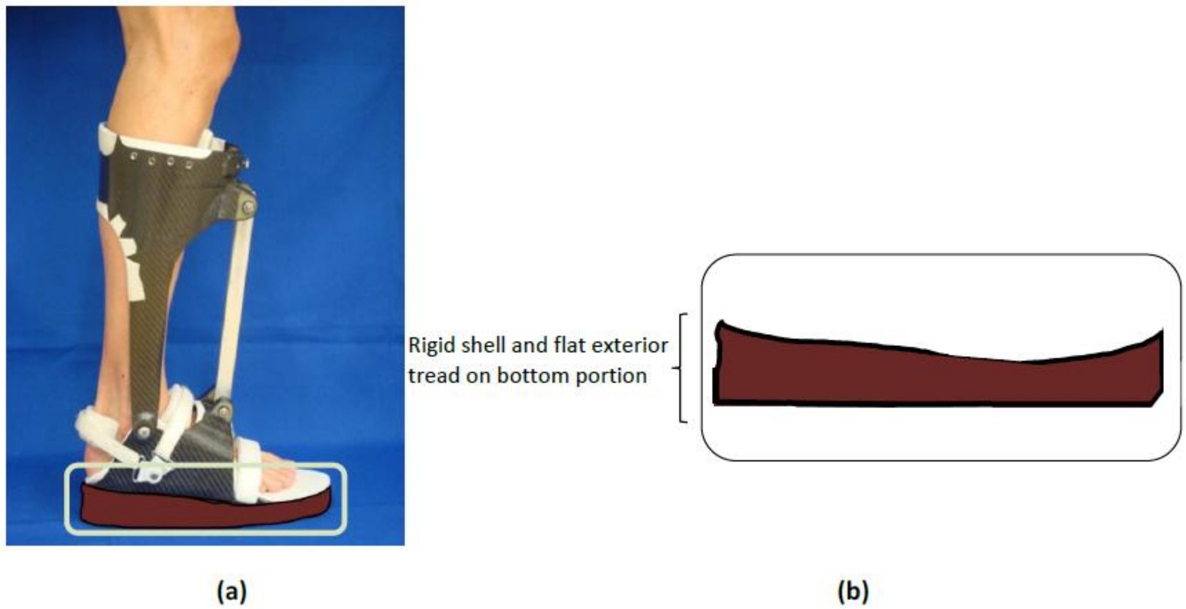


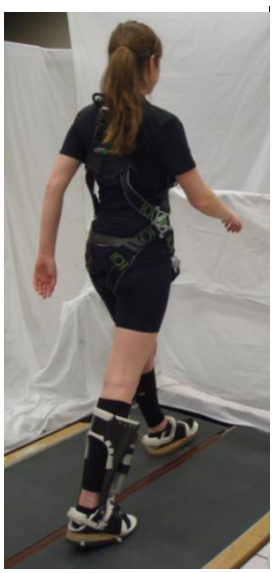
Figure 5. Components of the no rocker profile footwear condition.
(a) The no rocker profile footwear condition is anchored to the bottom portion of the foot shell in the STOP-NR condition. Footwear section represented as colored region. **(b)** Rigid carbon composite shell (brown) contains a flat exterior bottom portion with no rounded contour except at terminal forefoot. Height of heel and midfoot sections match height of rocker profile heel and midfoot sections. Tread is attached to the flat exterior bottom portion (black).

Motion capture

Sixteen passive reflective markers (14 mm) were attached to the skin of the lower limbs and pelvis of healthy subjects according to the Helen Hayes protocol (Kadaba, 1989). Markers at the shank and foot were attached to the exterior of the orthosis on the shank and foot cylinders because the orthosis prevented attaching markers to the skin in these regions. In addition to these sixteen markers, four additional markers were attached to the lateral sole of the footwear (two per feet) (**Figure 6b**). The first marker was placed posteriorly on the sole of the footwear and distal to the heel and the other was placed interiorly on the sole of the footwear and distal to the metatarsophalangeal joints. The purpose of these markers was to capture the landmarks which define early, middle and late stance as described by Perry (Perry 2010). Early stance is defined as heel strike to foot flat, middle stance is defined as foot flat to heel rise and late stance is defined as heel rise to toe off. For this purpose, it was important that the posterior and anterior sole maker were placed on a straight line to ensure that the landmarks were accurately determined.

Force and motion data were collected for each limb and synchronized as subjects walked on a custom dual belt instrumented treadmill containing force plates underneath each treadmill (1,080Hz, Advanced Mechanical Technology Inc., Watertown, MA) (Kram 1998; Toney 2013). Simultaneous motion data were captured using a 6-camera motion analysis system (120Hz, Vicon Motion Systems, Oxford, UK). All data were collected in the Vicon workstation and motion data were processed using the plug-in-gait model to identify and label markers. All data were imported to MATLAB version 7.11.0 (The Mathworks Inc., Natick, MA) for additional processing. Raw force signals were filtered (4th order Butterworth low pass filter with cutoff frequency of 20 Hz) and analyzed to determine ground reaction components and joint moments during stance and to identify stance and swing phases. Motion data were filtered (4th order Butterworth low pass filter with cutoff frequency of 10 Hz) and analyzed to determine motion of the ankle joint. All data were synchronized and time normalized to 100% of the gait cycle. Because the dominant motions of the ankle joint complex occur through the talocrural articulation as plantarflexion and dorsiflexion during gait, analysis of ankle motion was restricted to the sagittal plane (Perry J, 2010; Zatsiorsky V 1998).

(a)



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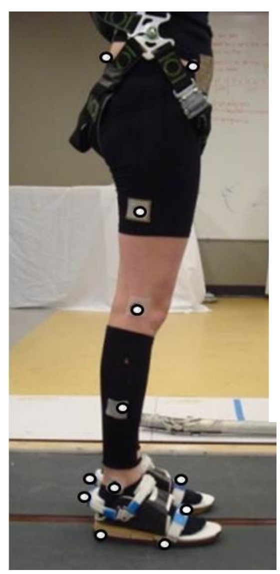
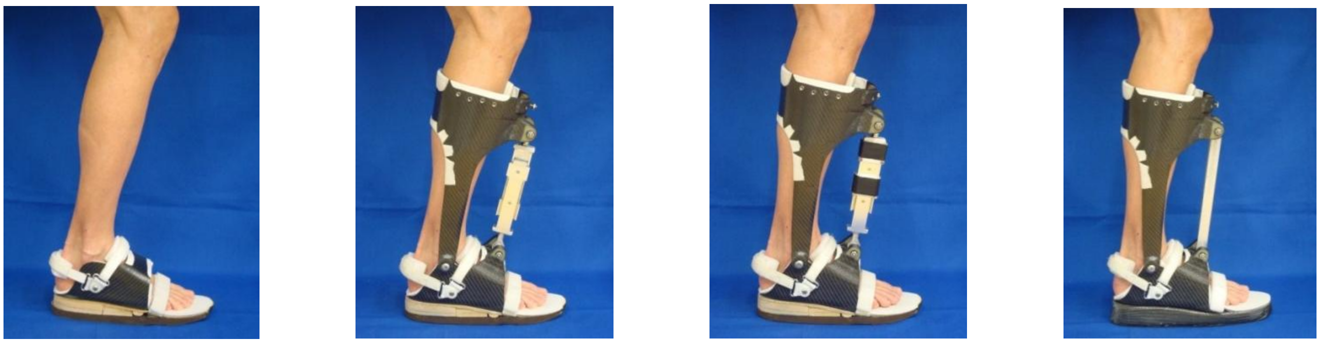


Figure 6. Experimental setup and reflective markers to create limb segment model.
(a) Subjects walked on a dual belt treadmill with force places beneath each treadmill. (Not pictured) Six high speed infrared cameras were used to capture the motion of the markers on the lower limbs of each subject. **(b)** Reflective markers on the rocker profile were used to render a linked segment model of each subjects's lower limb and capture motion during early, middle and late stance phases of gait.



a.) No AFO **b.) FREE** **c.) STOP** **d.) STOP -NR**

Figure 7. Conditions in the experimental protocol.
(a) No AFO condition allows free ankle plantarflexion and dorsiflexion, **(b)** FREE condition allows free ankle plantarflexion and dorsiflexion, **(c)** STOP condition constrains ankle plantarflexion and dorsiflexion, **(d)** STOP-NR condition constrains ankle plantarflexion and dorsiflexion and restrains foot motion. See **Table 1** for additional details.

Table 1. Condition components and motion controls.
Each of the four conditions: No AFO, FREE, STOP, and STOP-NR, describing the presence or absence of AFO and footwear components and the motion controls of each component. Note: Plantarflexion (PF), Dorsiflexion (DF).

	Components		
Condition	**AFO** **present (+) or absent (−)**	**Footwear** **present (+) or absent (−)**	**Motion Controls**
No AFO	−	+	Free Ankle PF and DF
FREE	+	+	Free Ankle PF and DF
STOP	+	+	Constrain Ankle PF and DF
STOP-NR	+	−	Constrain Ankle PF and DF Constrain Forefoot PF and DF

Data collection protocol

To quantify the effect of the rocker profile footwear on rollover in subjects using the custom AFO system during dynamic loading, individuals walked in four conditions: use of rocker profile footwear with no AFO (no AFO), use of rocker profile footwear and right unilateral AFO with no ipsilateral ankle constraint (FREE), use of rocker profile footwear with right unilateral AFO and maximal ipsilateral ankle constraint (STOP) and use of no rocker profile footwear with right unilateral AFO and maximal ipsilateral ankle constraint (STOP-NR) (**Figure 7, Table 1**). Rocker profiled footwear with no AFO was used on the contralateral leg in all walking conditions. The no AFO condition serves as the minimal constraining condition and is designed to mimic walking in shoes. Bilateral rocker profile footwear was used instead of shoes to control for the variety in shoe design, mass and size between different individuals. Due to the difference in the rocker profile design compared to shoes, the rollover outputs from with bilateral rocker profile footwear cannot be directly compared to those from different studies and were defined in this study. The FREE condition serves as the minimal constraining AFO condition which allows ankle-foot motion via the linear bearing. The FREE condition was designed to produce the same ankle-foot motion as the no AFO condition and serve as a more appropriate control condition. The STOP condition serves as the maximal constraining AFO condition which stops ankle-foot motion via blocks which restrict movement of the linear bearing. However, unlike the STOP-NR, the STOP condition includes the rocker profile footwear which is designed to simulate ankle-foot motion. Finally, the STOP-NR condition also serves as the maximal constraining AFO condition which includes the blocks to restrict the linear bearing movement and the rigid footwear to restrict fore-foot motion.

The data collection involved two lab visits. During the first visit, subjects completed a questionnaire and performed multiple trials of motor screening tasks (ball kick, stair ascent, standing balance recovery) (Lin W-H 2009, Zakas A 2006, Hoffman M 1998, Verhagen E 2005) to confirm their right lower limb was dominant, and were fit with custom designed footwear and a right ipsilateral AFO by a certified orthotist. In addition, subjects participated in a protocol to identify their preferred walking speed over ground and on a treadmill (Amorim 2009), and to be familiarized to the treadmill (Zeni 2010) prior to the motion capture portion of the protocol that occurred in a second visit. On the second visit, subjects walked at their preferred speed on a treadmill for three minutes in a randomized order of each experimental condition (**Figure 8**). To minimize carry over effects from one experimental condition to the next, we included a "wash out" period following each experimental condition. In the washout periods, subjects walked using bilateral rocker profile footwear (i.e. no AFO condition). Thus, the washout condition was designed to allow subjects to "de-adapt" from the effects of rollover (stance phase duration, cadence and radius of curvature) and energetic outputs (heart rate and rating of perceived exertion) influenced by the prior experimental condition and return to a "baseline" rollover consistent with the initial control condition (Bastian, 2008). Each condition was limited to three minutes due to the negative impacts that would have arisen from walking in the STOP-no rocker condition for an extended period. During the data collection, each subject's heart rate and rating of perceived exertion was recorded.

Initial No AFO	Rest	FREE	Rest	WO1 No AFO	Rest	STOP	Rest	WO2 No AFO	Rest	STOP-NR	Rest	WO3 No AFO
3 min	8 min	3 min	3 min	3 min	8 min	3 min	3 min	3 min	8 min	3 min	3 min	3 min

Figure 8. Representative sequence in experimental protocol.
Subjects walked using No AFO condition in initial period and three washout periods (initial No AFO, WO1 No AFO, WO2 No AFO, WO3 No AFO). Subjects also walked in three conditions using AFO (FREE, STOP, STOP-NR). A seated rest period (Rest) and washout period followed each condition. The order of each AFO condition was randomized for each subject.

Quantifying rollover

For this study, rollover will be quantified as stance phase duration, cadence, and radius of curvature. These outputs were chosen because they capture the result of lower limb rollover dynamics during stance (Oludare 2012). Stance phase duration quantifies the relative time of rollover dynamics, cadence quantifies the frequency of rollover, and radius of curvature quantifies the trajectory of the lower limb during stance.

Stance Phase Duration

The walking gait cycle is the period from initial contact (heel strike) to the next initial contact of the same limb and it lasts approximately one second (Perry 2010). The stance phase is typically normalized to total time of the gait cycle that also includes a swing phase. Normalized to the gait cycle, the stance phase occurs for 60% of the total gait cycle and swing phase occurs for 40% (Perry 2010, Lee 2008). For this study, and for previous literature, the stance phase is identified using the presence or absence of a ground reaction force as detected by the force plates.

Cadence

Cadence during walking is defined as the number of steps taken during a duration of walking and is typically expressed as steps/minute. In this study, cadence is reported as steps/minute. Typically, an able-bodied human is able to complete a gait cycle (or step) in one second, which would equate to 1 step/second or 60 steps/minute.

Radius of Curvature

The radius of curvature is a well-described rocker profile measurement which characterizes the movement of the lower limb over its center of pressure (Hansen 2000). The radius of curvature is based on the inverse pendulum model of walking which describes the human body in motion as a point mass and a stick that traverses space through collision and propulsion forces at the beginning and end of stance, respectively. In the inverse pendulum of walking model, the body's center of mass swings from collision to propulsion with pendulum like motion. It is this motion that the radius of curvature captures. To calculate the radius of curvature, the motion of the shank segment, tracked in the laboratory coordinate system, is tracked relative to the lower limb center of pressure in a shank-based coordinate system **(Figure 9)**.

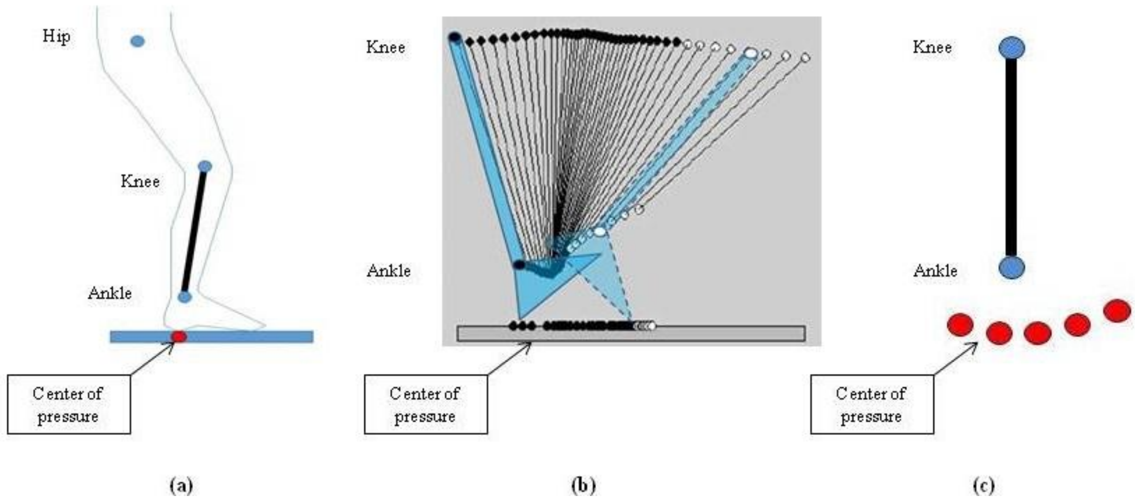


Figure 9. Laboratory and shank coordinate systems during stance phase.
(a) Hip, knee and ankle markers (solid blue circle) and location of center of pressure under the foot (solid red circle) in a laboratory-based coordinate system.
(c) The movement of the lower limb in the laboratory-based coordinate system. The lower limb moves from early stance to late stance phase. In early stance the shank (solid blue rectangle) is reclined the ankle-foot (solid blue triangle) complex is in dorsiflexion. In late stance, the shank (dashed blue rectangle) is inclined and the ankle-foot complex (dashed blue triangle) is in plantarflexion.
(b) Knee and ankle markers (solid blue circle) and location of center of pressure under the foot (solid red circles) in a shank based coordinate system.

Component analysis of rocker profile footwear

Periods of stance phase duration

The stance phase can be divided into three major periods: early, middle and late stance phase. The durations of these periods of stance has been reported extensively by Perry. Early stance is 0-12% of the gait cycle, middle stance is 12-30%, late stance is 30-50% and pre-swing is 50-62% **(Figure 10)**. For this study, late stance and pre-swing have been combined as late stance. To quantify the duration of each period, a custom marker set was used to track the alignment of the ankle-foot complex **(Figure 11).** Early stance was quantified as the period from initial contact to foot flat, middle stance was quantified as the period from foot flat to heel rise, and late stance was quantified as the period from heel rise to toe off.

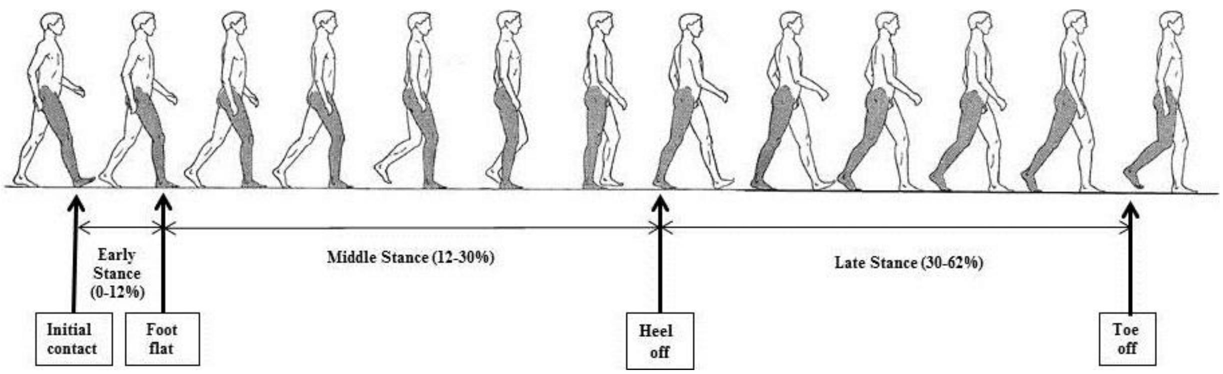


Figure 10. Stance phase periods of the gait cycle.
Right leg depicts early stance period (between initial contact and foot flat [0-12% of gait cycle]), middle stance period (between foot flat and heel off [12-30% of gait cycle]), and late stance period (between heel off and toe off [30-62% of gait cycle]). Image modified from: Ducroquet R, Ducroquet J, Ducroquet P (eds.) Walking and limping a study of normal and pathological walking. Philadelphia: LB Lippincott Company; 1968, p. 18.

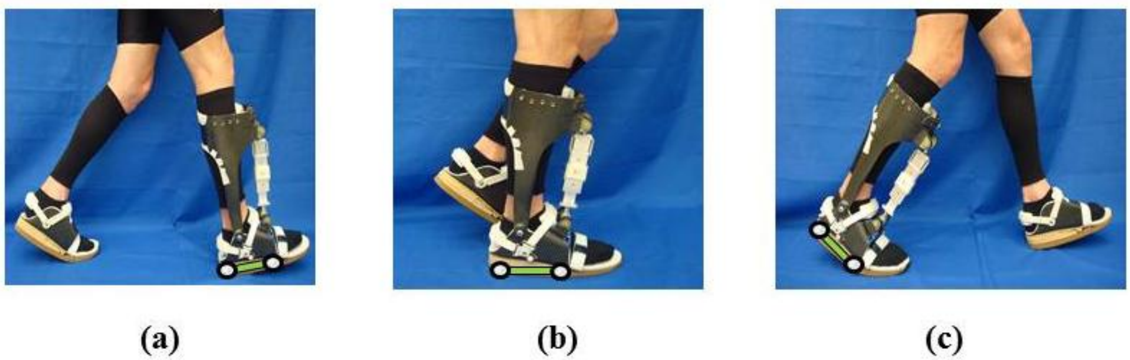


Figure 11. Quantifying period of stance phase durations
Two markers (white circles with black outline) embedded into the rocker profile footwear were used to capture the alignment of the ankle-foot complex. The alignment of the ankle-foot complex was used to determine and quantify the periods of stance phase **(a)** early stance was quantified as the period between heel strike and foot flat, **(b)** mid stance was quantified as the period of foot flat to heel rise, and **(c)** quantified as the period of heel rise to toe off.

Ground reaction forces

The ground reaction force is the resultant force that arises when the lower limb is in contact with the ground. For the purposes of this study, the ground reaction force has been decomposed into the vertical and horizontal components. As discussed earlier in the introduction, there are three significant periods of high force magnitude during the gait cycle. The first period is initial contact to loading response, the second is loading response (where the force is roughly equal to body weight), and the third is heel rise to toe off. The magnitude and timing of both the vertical and horizontal components of the ground reaction forces during the first, second and third periods have been well characterized and reported by many previous investigators (Lee 2008).

Energetic outputs during rollover

For this study, energy expenditure will be quantified as heart rate and rating of perceived exertion. These outputs were chosen because they capture metabolic activity during walking. Heart rate quantifies the metabolic effort as the output by the cardiovascular system. Rating of perceived exertion quantifies the perceived effort by the individual. Each of this outputs are commonly used and are well described (Tonis TM 2012, Chen MJ 2002).

Heart rate

The heart rate is a measure of the metabolic energy expended by an individual. For non-athletic individuals at rest, their heart rate is typically between 70 and 90 beats per minute. During vigorous activity, the heart rate increases to supply more oxygenated blood to the muscular system to account for the increased consumption of energy. For this study, the heart rate is expressed as the number of heart beats per minute and serves as an indirect measure of energy expenditure.

Rating of perceived exertion

The rating of perceived exertion (RPE) is a subjective measure on a scale of 6 to 20 which quantifies the individual's reported effort during an activity (i.e. running, walking, and cycling). The lowest score, 6, indicates no effort and the highest score, 20, indicates maximal effort.

Statistical analysis

The change in all subjects' (n=4) mean rollover and energy expenditure outputs for each condition during minute three (final minute of walking for each condition) were statistically analyzed. The analyses were performed on 60 seconds of the third minute because in that minute all subjects achieved the least variability in rollover outputs which was defined as steady state. All statistical analysis was performed with SPSS version 21 (SPSS IBM, New York, U.S.A).

For the first hypothesis, the change in the mean of all subjects' (n=4) rollover and energy expenditure outputs in the no AFO condition and the FREE condition are compared using a two tailed paired T-test (α=0.05). For the second hypothesis, the change in mean of all subjects' (n=4) rollover and energy expenditure outputs in the FREE condition and the STOP-NR condition are compared using a two tailed paired T-test (α=0.05). For the third hypothesis, the change in the mean of all subjects' (n=4) rollover and energy expenditure outputs in the STOP condition are compared to the FREE and the STOP-NR condition using repeated measures ANOVA with a Bonferroni post hoc analysis was used to examine the differences.

As an additional assessment, we examined the change in mean duration of stance and mean ground reaction forces during early, middle and late stance periods and related these findings to define the rollover performance of the rocker profile footwear sections (cushion heel, midfoot platform, forefoot rocker) during FREE, STOP and STOP-NR conditions. These differences were examined using repeated measures ANOVA with a Bonferroni post hoc analysis. A 95% confidence interval of the FREE condition was also used to compare the changes in the FREE condition to the STOP and STOP-NR conditions.

CHAPTER 3

RESULTS

Defining rollover

No AFO condition: initial period compared to washout periods

There was no statistically significant difference ($p>0.05$) between the mean rollover and mean energy expenditure outputs of subjects using the no AFO condition in the initial period and the three washout periods. The mean rollover outputs using the no AFO condition were compared between the initial period and three washout periods to determine if there were any carryover effects in rollover outputs accumulated in the previous experimental condition that were transferred to the rollover outputs accumulated in the washout periods (**Table 2**).

Table 2. No AFO condition – initial period compared to washout periods
Ipsilateral rollover outputs mean (SD) and energetic outputs mean (SD) during minute 3 for all subjects (n=4) using the no AFO condition in initial period and three washout periods. All values not significant ($p>0.05$).

	Periods using No AFO Condition			
Rollover outputs	**Initial**	**Washout1**	**Washout2**	**Washout3**
Stance phase duration (% gait cycle)	59.4 (0.7)	61.4 (0.9)	61.2 (0.8)	60.9 (0.7)
Cadence (steps·min^{-1})	55.8 (2.8)	54.7 (3.1)	54.3 (2.3)	55.3 (2.8)
Radius of curvature	0.28 (0.03)	0.27 (0.01)	0.27 (0.01)	0.28 (0.01)
Energetic outputs	**Initial**	**Washout1**	**Washout2**	**Washout3**
Heart rate (beats·min^{-1})	103.8 (10.3)	110.0 (11.3)	108.0 (13.5)	111.0 (7.7)
Rating of Perceived Exertion	9.0 (0.8)	9.3 (1.5)	9.3 (1.5)	9.0 (1.4)

Comparison of rollover and energy expenditure between no AFO and FREE conditions

The mean rollover outputs were compared between the no AFO and the FREE conditions to determine if the FREE condition would be represented as the control condition. There was no statistically significant difference ($p>0.05$) between the mean rollover and mean energy expenditure outputs of all subjects using the no AFO condition and FREE condition except for significant difference ($p<0.05$) in mean radius of curvature and mean heart rate. The mean radius of curvature and the mean heart rate in the FREE condition were significantly ($p<0.05$) greater than in the no AFO condition (**Table 3**).

Table 3. Comparison of no AFO (initial) and FREE conditions

Ipsilateral rollover outputs mean (SD) and energetic outputs mean (SD) during minute 3 for all subjects (n=4) in no AFO and FREE conditions. * Significance ($p<0.05$).

	Conditions	
Rollover outputs	**No AFO**	**FREE**
Stance phase duration (% gait cycle)	59.4 (0.7)	59.5 (0.8)
Cadence (steps·min^{-1})	55.8 (2.8)	54.0 (2.6)
Radius of curvature	0.28 (0.03)	0.29 (0.01)*
Energetic outputs	**No AFO**	**FREE**
Heart rate (beats·min^{-1})	103.8 (10.3)	110.5 (6.7)*
Rating of Perceived Exertion	9.0 (0.8)	9.0 (1.4)

Comparison of rollover and energy expenditure between FREE and STOP-NR conditions

The mean rollover outputs of subjects in the STOP-NR condition were compared to the FREE condition to quantify the effective range of rollover performance. There was no statistically significant difference ($p>0.05$) in the mean rollover and mean energy expenditure of subjects in the FREE and STOP-NR conditions except for significant difference ($p<0.05$) of subjects' radius of curvature and heart rate. Radius of curvature and heart rate of subjects in the STOP-NR condition was significantly ($p<0.05$) greater compared to the FREE condition. Subjects in STOP-NR condition elicited a significant difference ($p<0.05$) that consisted of smaller mean stance phase duration, larger mean cadence, larger radius of curvature and increased energy expenditure compared to the FREE condition **(Table 4).**

Table 4. Comparison of FREE and STOP-NR conditions

Ipsilateral rollover outputs mean (SD) and energetic outputs mean (SD) during minute 3 for all subjects (n=4) in FREE and STOP-NR conditions. # Significance ($p<0.05$).

	Conditions	
Rollover outputs	**FREE**	**STOP-NR**
Stance phase duration (% gait cycle)	59.5 (0.8)	56.9 (0.2)
Cadence (steps·min^{-1})	54.0 (2.6)	57.0 (6.1)
Radius of curvature	0.29 (0.01)	0.69 (0.12) #
Energetic outputs	**FREE**	**STOP-NR**
Heart rate (beats·min^{-1})	110.5 (6.7)	131.5 (8.1) #
Rating of Perceived Exertion	9.0 (1.4)	12.0 (0.8)

Comparison of rollover and energy expenditure between FREE, STOP and STOP-NR conditions

The mean rollover outputs of subjects in the STOP condition were compared to the FREE and STOP-NR conditions to quantify the effective range of rollover performance. There was no difference ($p>0.05$) in subjects mean rollover and mean energy expenditure outputs in the STOP condition compared to the FREE and STOP-NR conditions except for a difference ($p<0.05$) in subjects' mean heart rate. Mean heart rate in the STOP condition was significantly lower ($p<0.05$) than in the STOP-no rocker condition but not different ($p>0.05$) in the FREE condition. Subjects elicited lower mean stance phase duration in the STOP condition compared to the FREE condition but higher stance phase duration in FREE condition compared to the STOP-NR condition. Mean cadence of subjects was lower in STOP compared to FREE and STOP-NR. Subjects elicited a significantly ($p<0.05$) larger mean radius of curvature in STOP-NR compared to FREE condition and non-significant ($p>0.05$) smaller mean radius of curvature in FREE compared to STOP-NR condition **(Table 5)**. Regarding subjects' mean energetic outputs, there was significantly ($p<0.05$) greater mean heart rate in the STOP-NR condition compared to the FREE condition but a non-significant ($p>0.05$) lower heart rate in STOP condition compared to FREE. Rating of perceived exertion was not significantly ($p>0.05$) greater in STOP-NR compared to FREE condition and STOP conditions **(Table 5).**

Table 5. Comparison of FREE, STOP and STOP-NR conditions
Ipsilateral rollover outputs mean (SD) and energetic outputs mean (SD) during minute 3 for all subjects (n=4) in FREE, STOP and STOP-NR conditions. # Significance for FREE and STOP-NR ($p<0.05$).
ɫ Significance for STOP and STOP-NR ($p<0.05$)

	Conditions		
Rollover outputs	**FREE**	**STOP**	**STOP-NR**
Stance phase duration (% gait cycle)	59.5 (0.8)	58.2 (0.1)	56.9 (0.2)
Cadence (steps·min^{-1})	54.0 (2.6)	53.3 (2.5)	57.0 (6.1)
Radius of curvature	0.29 (0.01)	0.34 (0.04)	0.69 (0.12) **#**
Energetic outputs	**FREE**	**STOP**	**STOP-NR**
Heart rate (beats·min^{-1})	110.5 (6.7)	111.3 (8.3)	131.5 (8.1) **# ɫ**
Rating of Perceived Exertion	9.0 (1.4)	10.5 (1.0)	12.0 (0.8)

Component analysis of rocker profile footwear

Comparison of stance period duration in FREE, STOP and STOP-NR

There was no significance difference ($p>0.05$) in the mean duration of the early, middle and late stance periods of subject's contralateral limb in the three conditions (FREE, STOP and STOP-NR). Also, there was no significant difference ($p>0.05$) in the duration of early, middle and late stance periods for the ipsilateral limb during each condition (FREE STOP, STOP-NR). On the ipsilateral limb, early stance duration in the STOP and STOP-NR conditions was longer than the FREE condition. Also in the ipsilateral limb, middle stance phase duration in STOP and in STOP-NR conditions was shorter than the in the FREE condition.

Table 6. Comparison of ipsilateral leg periods of stance duration in FREE, STOP and STOP-NR
Periods of stance duration outputs mean (SD) in FREE, STOP and STOP-NR during minute 3 for all subjects (n=4). All values not significant (p>0.05)

	Conditions		
Period of stance (% gait cycle)	FREE	STOP	STOP-no rocker
Early stance	7.7 (1.3)	15.0 (3.7)	13.5 (2.6)
Middle stance	37.4 (0.9)	25.4 (3.3)	24.5 (4.6)
Late stance	19.5 (1.5)	21.8 (4.2)	21.9 (5.1)

Table 7. Comparison of contralateral leg periods of stance duration in FREE, STOP and STOP-NR
Periods of stance duration outputs mean (SD) in FREE, STOP and STOP-NR during minute 3 for all subjects (n=4). All values not significant (p>0.05)

	Conditions		
Period of stance (% gait cycle)	FREE	STOP	STOP-no rocker
Early stance	9.2 (1.3)	8.9 (1.6)	8.34 (1.7)
Middle stance	33.0 (2.7)	33.0 (2.7)	28.8 (3.4)
Late stance	21.4 (2.8)	22.1 (4.2)	23.0 (4.4)

Comparison of ground reaction forces in FREE, STOP and STOP-NR

During early stance and late stance periods, the STOP-no rocker ipsilateral limb horizontal and vertical components of the ground reaction forces have two peaks as opposed to the single peaks of the STOP and FREE conditions. For the ipsilateral limb, both components of the ground reaction force magnitude and timing were statistically different ($p<0.05$) in early, mid and late stance periods for the STOP condition compared to the STOP-no rocker but not statistically different ($p>0.05$) for the STOP condition compared to the FREE condition. For the horizontal component of the ground reaction force in early stance, the peak force in the STOP-no rocker condition is statistically significantly lower ($p<0.05$) and earlier ($p<0.05$) than the peak forces of the STOP and FREE condition. Also, in the late stance period, the peak force of the horizontal component of the ground reaction force is significantly lower ($p<0.05$) and later ($p<0.05$) in the STOP-no rocker condition compared to the STOP and FREE conditions. For the vertical component of the ground reaction force in early stance, the peak force in the STOP-no rocker condition is significantly later ($p<0.05$) than the peak forces of the STOP and FREE condition but of similar magnitude ($p>0.05$). Also, in late stance period, the peak force of the vertical component of the ground reaction force is significantly lower and earlier in the STOP-no rocker condition compared to the STOP and FREE conditions.

There was no significant difference ($p>0.05$) in the peak forces magnitude or timing for the horizontal and vertical ground reaction forces on the ipsilateral limb during the FREE, STOP and STOP-NR except for the timing of the horizontal force in early stance and the magnitude of the horizontal force in late stance. The timing of the peak horizontal force in the STOP-NR was significantly later ($p<0.05$) compared to FREE and STOP. The magnitude of the peak horizontal force in STOP-NR was significantly lower ($p<0.05$) compared to FREE and STOP.

The only significant difference in the contralateral limb ground reaction forces are in the middle stance period of the horizontal ground reaction force where the force in the STOP-no rocker condition is significantly greater ($p<0.05$) compared to the FREE and STOP conditions. There was no statistically significant difference ($p<0.05$) in the peak forces or timing for the timing for the horizontal and vertical ground reaction forces on the ipsilateral limb during the FREE, STOP and STOP-NR.

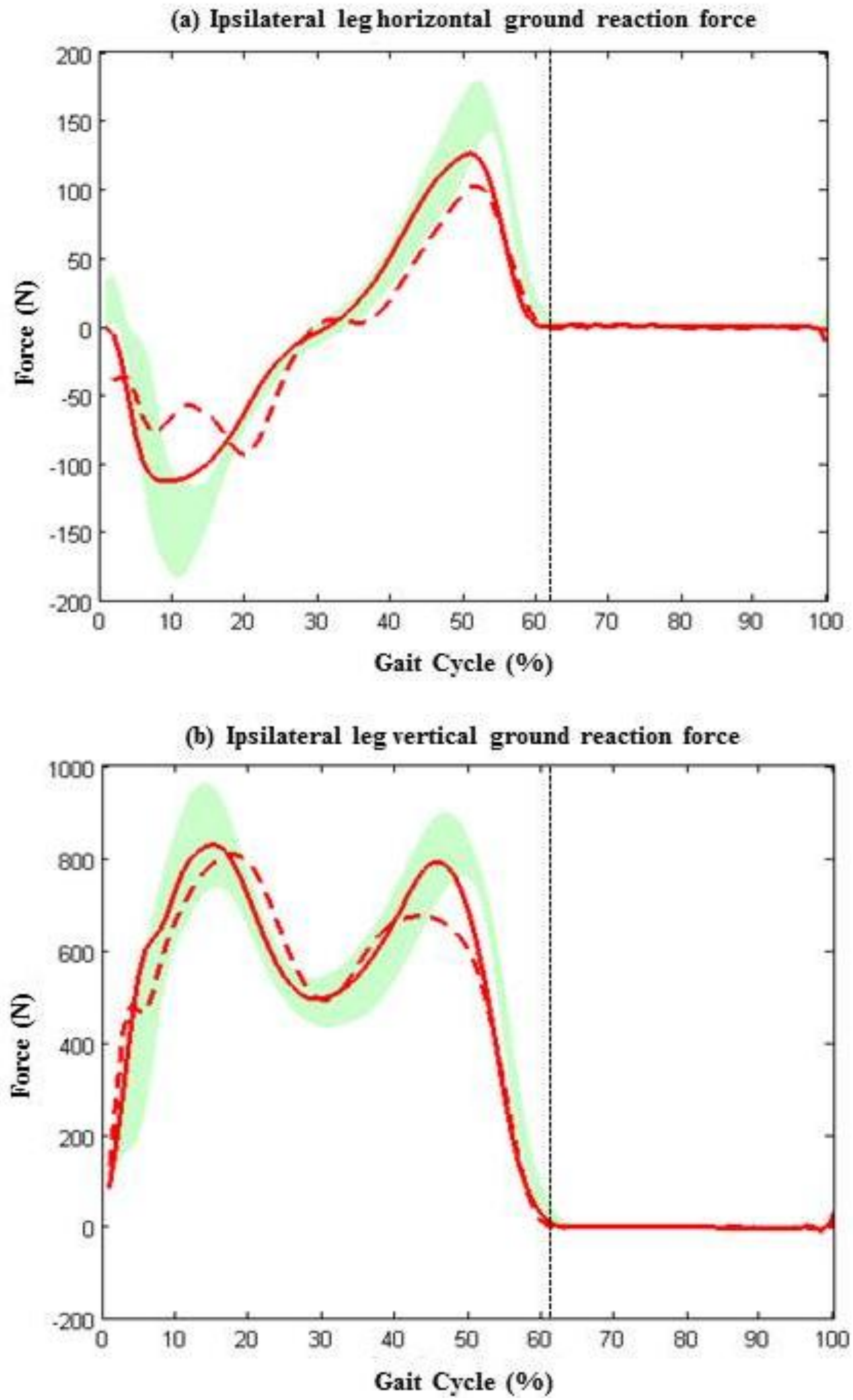


Figure 12. Comparison of ipsilateral leg ground reaction forces in AFO conditions
(a) Horizontal ground reaction force (N) normalized to gait cycle (%) in FREE, STOP and STOP-NR during minute 3 **(b)** Vertical ground reaction force (N) normalized to gait cycle (%) in FREE, STOP and STOP-NR during minute 3.
95 confidence interval of FREE (shaded green region), mean of STOP (red solid line), and mean of STOP-NR (red dotted line) for all subjects (n=4). Toe off in FREE (vertical dashed line)

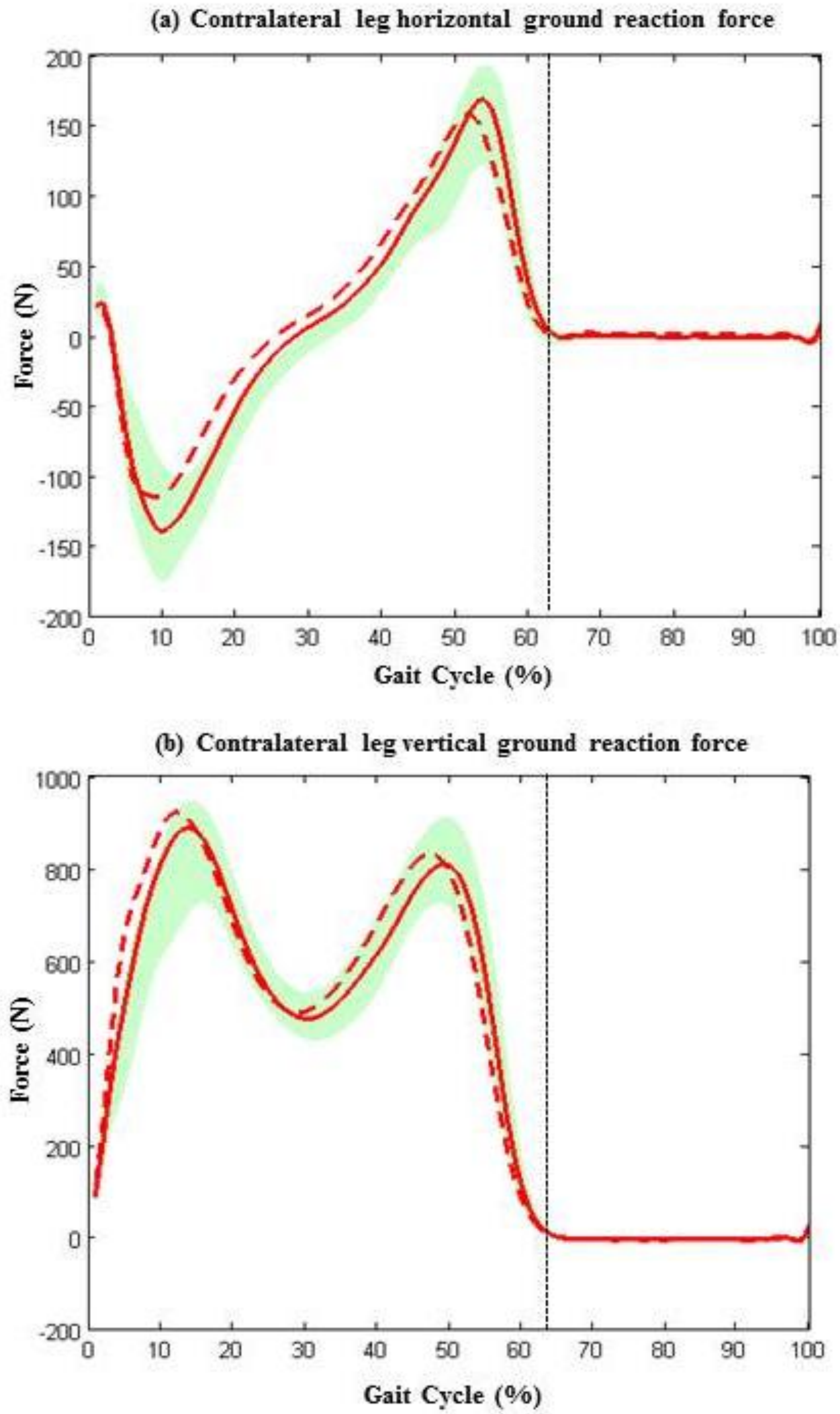


Figure 13. Comparison of contralateral leg ground reaction forces in AFO conditions
(a) Horizontal ground reaction force (N) normalized to gait cycle (%) in FREE, STOP and STOP-NR during minute 3 **(b)** Vertical ground reaction force (N) normalized to gait cycle (%) in FREE, STOP and STOP-NR during minute 3.
95 confidence interval of FREE (shaded green region), mean of STOP (red solid line), and mean of STOP-NR (red dotted line) for all subjects (n=4). Toe off in FREE (vertical dashed line)

Table 8. Comparison of ipsilateral leg early stance ground peak reaction forces.
Peak ground reaction force (GRF) mean (SD) during minute 3 for all subjects (n=3) in FREE, STOP and STOP-NR conditions. *FREE-STOP-NR significant (p<0.05), #FREE-STOP significant (p<0.05), ǂSTOP-STOP-NR significant (p<0.05)

		Conditions			
		FREE	STOP	STOP-NR	
Horizontal GRF	Peak (N)	156.8 (38.5)	125.1 (29.7)	70.6 (41.4)	148.3 (45.5)
	% gait cycle	11.0 (0.0)	9.0 (1.0)	7.3 (0.6) *	15.0 (6.6)
Vertical GRF	Peak (N)	871.0 (133.4)	853.1 (146.3)	506.0 (54.6) #	922.3 (174.6)
	% gait cycle	14.3 (0.6)	15.7 (0.6)	4.3 (1.3) #ǂ	16.7 (4.0)

Table 9. Comparison of ipsilateral leg late stance peak ground reaction forces
Peak ground reaction force (GRF) mean (SD) during minute 3 for all subjects (n=3) in FREE, STOP and STOP-NR conditions. *FREE-STOP-NR significant (p<0.05)

		Conditions		
		FREE	STOP	STOP-NR
Horizontal GRF	Peak (N)	168.7 (17.0)	135.8 (15.4)	100.7 (8.9) *
	% gait cycle	52.3 (1.2)	50.3 (0.6)	53.3 (2.1)
Vertical GRF	Peak (N)	849.6 (76.8)	840.0 (90.3)	759.9 (45.9)
	% gait cycle	47.7 (1.5)	45.5 (0.6)	42.0 (2.6)

Table 10. Comparison of ipsilateral leg mid stance peak ground reaction forces.
Peak ground reaction force (GRF) mean (standard deviation) during minute 3 for all subjects (n=3) in FREE, STOP and STOP-NR conditions. All conditions not significant (p>0.05)

		Conditions		
		FREE	STOP	STOP-NR
Vertical GRF	Peak (N)	503.2 (43.9)	513.8 (27.7)	507.1 (53.8)
	% gait cycle	28.7 (1.5)	30.0 (1.0)	30.0 (2.0)

Table 11. Comparison of contralateral leg early stance peak ground reaction forces.
Peak ground reaction force (GRF) mean (standard deviation) during minute 3 for all subjects (n=4) in FREE, STOP and STOP-NR conditions. All conditions not significant (p>0.05)

		FREE	STOP	STOP-NR
Horizontal GRF	Peak (N)	138.4 (37.4)	142.0 (32.5)	116.5 (47.7)
	% gait cycle	11.0 (2.0)	10.3 (1.3)	8.8 (1.7)
Vertical GRF	Peak (N)	835.5 (113.7)	889.0 (125.7)	923.3 (164.1)
	% gait cycle	15.3 (1.0)	13.5 (1.3)	12.8 (1.3)

Table 12. Comparison of contralateral leg late stance peak ground reaction forces.
Peak ground reaction force (GRF) mean (standard deviation) during minute 3 for all subjects (n=4) in FREE, STOP, and STOP-NR conditions. All conditions not significant (p>0.05)

		FREE	STOP	STOP-NR
Horizontal GRF	Peak (N)	161.0 (32.5)	170.1 (27.5)	161.9 (25.0)
	% gait cycle	54.3 (1.3)	53.8 (1.0)	52.0 (1.2)
Vertical GRF	Peak (N)	822.4 (90.1)	808.8 (77.9)	829.6 (75.0)
	% gait cycle	49.3 (1.0)	49.3 (1.0)	47.8 (1.0)

Table 13. Comparison of contralateral leg mid stance peak ground reaction forces
Peak ground reaction force (GRF) mean (standard deviation) during minute 3 for all subjects (n=4) in FREE, STOP, and STOP-NR conditions. All conditions not significant (p>0.05)

		FREE	STOP	STOP-NR
Vertical GRF	Peak (N)	477.4 (54.1)	468.7 (67.3)	485.9 (104.6)
	% gait cycle	30.8 (1.0)	29.5 (2.1)	29.3 (1.0)

CHAPTER 4

DISCUSSION

FREE condition characterizes effective rollover and STOP-NR characterizes ineffective rollover

In this study, the mean rollover outputs of subjects walking on a treadmill during minute 3 in the no AFO condition defined a baseline of high performance rollover. The mean rollover outputs were mean stance phase duration, mean cadence, mean radius of curvature. To further characterize rollover, we included the mean energetic outputs which consisted of subjects' mean heart rate and mean rating of perceived exertion. The results in the no AFO rollover outputs appear in **Table 2**. To determine whether the FREE condition was an acceptable Control condition (i.e., provided effective rollover), the mean rollover outputs of subjects in the FREE condition were compared to the subjects' rollover outputs in the no AFO condition. Results revealed there was no significant difference ($p>0.05$) in rollover outputs (mean stance phase duration, mean cadence, mean radius of curvature) elicited by subjects in the no AFO condition and FREE conditions except for significant difference ($p<0.05$) in mean radius of curvature and mean heart rate. However subjects' radius of curvature and heart rate in the FREE condition was 3.6% and 6.8% greater than in the no AFO condition respectively. Based on these results, we accepted hypothesis 1 – meaning that rollover outputs in the FREE condition are not different compared to the no AFO condition. Thus, the FREE condition provides effective rollover which makes it appropriate to serve as the control condition. As such, the FREE condition is an appropriate substitute for the no AFO condition because it does not significantly change subjects' rollover performance.

Regarding rollover performance, results revealed that subjects in the STOP-NR condition performed less effectively than in the FREE condition. This finding provides important new evidence that maximum rigidity of materials (i.e., carbon fiber) and flat shape (i.e., no rounded contour) in a footwear design appear to notably compromise roll over. While this is somewhat intuitive, no previous evidence appears to have been reported. In addition, subjects walking in the STOP-NR condition adopted different movement patterns in response to the high rigidity and flat shape of this type of footwear system. In particular, one of the subjects adopted a pointed toe gait which was drastically different from the abrupt heel to toe gait adopted by the other subjects. Due to this notable difference in movement strategies to the rigid footwear system, the subject's that adopted the drastically different rollover outputs in STOP-NR was not included in the analysis of rollover performance analysis. In the STOP-NR condition, rollover was characterized by shorter stance phase duration, higher cadence, larger radius of curvature (i.e., flatter, less rounded shape). Energetics were characterized by higher heart rate and higher rating of perceived exertion in STOP-NR than in the FREE condition. However, only the radius of curvature and heart rate were significantly different between the FREE and STOP-NR conditions. There was a 138% increase in radius of curvature and a 19% increase in heart rate by subjects in the STOP-NR condition compared to the FREE condition. Based on these results, we accepted hypothesis 2 which states that the rollover performance of subjects in the STOP-NR condition will be different than in the FREE condition. It can also be concluded that rollover performance characterized in the STOP-NR condition is ineffective due to and the larger radius of curvature (i.e., flatter, less rounded radius) and. the increased energy expenditure elicited by subjects.

Rocker profile footwear provides effective rollover

After determining that the FREE condition was characteristic of effective rollover and that the STOP-NR condition was characteristic of ineffective rollover, we were able to create a metric to determine the rollover performance of the STOP condition. Establishing a metric to determine rollover performance of the STOP condition was important because the majority of lower limb orthosis systems function to restrain ankle motion. Due to orthotic ankle motion constraint, footwear is combined with an ankle foot orthosis to enable the wearer to achieve minimum disruption in roll over. Thus, examining the roll over performance achieved by subjects in the STOP condition would provide important insights for designers of footwear and clinicians that fit footwear to persons that use lower limb orthoses as a motion control treatment.

Results revealed there were no significant differences ($p>0.05$) in the rollover outputs of subjects walking in the STOP condition compared to the FREE and STOP-NR conditions with the exception of heart rate, which was significantly ($p<0.05$) lower in the STOP condition compared to the STOP-NR condition. Subjects elicited mean heart rate that was essentially no different (i.e., was 0.01% greater in the STOP compared to the FREE condition), but was 15.1% lower in the STOP compared to the STOP-NR. Although the radius of curvature and heart rate of subject were not significantly different in the STOP compared to the STOP-NR conditions, there was a 102.9% increase in the radius of curvature and a 18.0% increase in the heart rate by subjects walking in the STOP-NR condition compared to the STOP condition. Based on these results, we partially accepted hypothesis 3 which states that rollover is no different in the STOP condition than in the FREE condition but is different from the STOP-NR condition. No definite conclusions can be drawn from the results comparing roll over in the STOP condition to the STOP-NR condition.

STOP condition does not alter braking and propulsive forces

In addition to examining rollover outputs of subjects walking in rocker profile footwear, the performance of each of the three sections of the rocker profile footwear components were specifically examined to quantify the influence of each component on rollover performance. The footwear components were analyzed by comparing subjects' horizontal and vertical ground reaction forces, and subjects' duration of early, middle and late stance phase walking in FREE, STOP and STOP-NR conditions. During early stance, subjects' horizontal and vertical ground reaction forces were representative of the body's braking force, whereas during late stance, the horizontal and vertical ground reaction forces are representative of the subjects' lower limb propulsive force. In the STOP condition compared to the FREE condition, there was no statistical difference in the magnitude or time in the gait cycle of the braking and propulsive ground reaction forces. Thus there was a preservation of ground reaction forces by subjects in the STOP condition indicating that during early stance, the heel portion of the rocker profile was effective in absorbing the force of the falling limb. Furthermore, we interpret these findings that the forefoot and toe portions of the rocker profile footwear did not impede propulsion during late stance. In the STOP-NR condition of subjects' ipsilateral limb, there was a reduction in the peak braking and propulsive forces which occur during early and late stance phase, respectively, compared to the FREE and STOP conditions. For the STOP-NR condition, the timing of the peak braking and propulsive forces also occurred earlier in early stance compared to the FREE and STOP conditions. The decrease in the braking and propulsive forces in the STOP-NR condition is likely attributed to the lack of ankle-foot motion in the lower limb due to the maximal constraint of ankle motion due to the orthosis and the maximum rigidity and flat shape of the rigid footwear. Because rollover was still achieved by subjects in the STOP-NR condition, we speculate that the knee and hip joints, and the lower limb musculature may be providing additional force to compensate for the force lost due to no ankle-foot motion. To confirm this speculation, further analysis of the knee and hip joint forces and moments, and the overall mechanical work performed by the body will need to be performed.

STOP and STOP-NR conditions increase early stance duration and decrease middle stance duration

The analysis of early, middle and late stance durations revealed that the there was no statistically significant difference between the FREE, STOP and STOP-NR conditions or between the ipsilateral and contralateral limbs. For the ipsilateral limb in the STOP and STOP-NR conditions however, there was a considerable increase in STOP and decrease in STOP-NR conditions in early and middle stance phase, respectively, compared to the FREE condition. In the STOP condition compared to the FREE condition, early stance duration was 48.7% greater and middle stance duration was 32.1% lower. In the STOP-NR condition compared to the FREE condition, early stance duration was 43.0% greater and middle stance duration was 34.5% lower. Also, in the STOP and STOP-NR conditions the differences were less than 5% during early and middle stance. The increase in early stance duration can be attributed to the lack of ankle plantarflexion during early stance in the STOP and STOP-NR conditions. In FREE condition, the ankle-foot complex undergoes plantarflexion during early stance phase which allows the foot to become lowered to the ground, but because plantarflexion motion has been constrained in both the STOP and STOP-NR conditions, the foot flat portion of early stance occurs later in the gait cycle (i.e., it is delayed). And due to early stance occurring for a longer duration, the duration of the middle stance period is shortened to preserve typical overall stance phase duration. Thus, the analysis of early, middle and late stance periods reveals that although, similar total stance phase duration is achieved in the FREE, STOP and STOP-NR conditions, it is not achieved distributed similarly within early, middle and late stance periods. Further research will need to be performed examine design changes in the rocker profile footwear that preserve the duration of stance phase periods during rollover.

Washout periods effectively eliminated adapted behavior

Based on the results revealing no difference in rollover performance of subjects using the no AFO condition during the initial period compared to the three washout periods that followed each experimental condition, we interpret this to mean that the washouts were effective in eliminating or “de-adapting” the movements that were shaped in each prior experimental condition. Subjects’ rollover outputs in each of the washout periods returned to the initial rollover output baseline established in the initial period of the no AFO condition. Thus, by effectively washing out the adapted movements and randomizing the data collection order of conditions, we were able to ensure that there was no order effect for each subject to each of the experimental conditions (i.e., FREE, STOP, STOP-NR). This provided greater confidence in our results because the likelihood of “contamination” in the outputs was minimized and likely subjets’ rollover outputs in response to each experimental condition were real and were not confounded by carry over effects.

Conclusions

Although the results from this study do not provide a clear conclusive decision on the efficacy of the rocker profile footwear in producing effective rollover, preliminary results revealed in this study enabled us to quantify rollover performance variables, to characterize the function of components in rocker profile footwear and to create a new experimental paradigm for studying the performance of rocker profile footwear. By examining the rollover outputs of subjects in the no AFO, FREE and STOP-NR conditions, we were able to create a metric that quantifies the performance range of the footwear in producing rollover. Until now, this type of analysis has not been performed, nor has any prior analysis revealed characteristics of rollover performance to define ineffective and effective rollover in footwear systems. By defining the range of rollover performance in a custom rocker profile footwear system, we were also able to suggest features that may enhance the efficacy of the rocker profile footwear relative to an effective and ineffective rollover. To confirm these novel findings, a larger sample size will be needed to confirm results from the experimental paradigm. In addition to creating the metric that defines a range of rollover performance of footwear, analyzing the ground reaction forces and the period of stance durations, enabled us to further determine how each component of the rocker profile footwear contributed to rollover outputs. Quantifying early, middle and late stance duration was also a novel method of analyzing rollover outputs which has the potential to be used as a metric in future studies aimed at optimizing the rollover performance and function of rocker profile footwear. Additionally, through this study we discovered that the radius of curvature and heart rate were the most sensitive parameters in defining rollover. As a result, we suggest that these parameters be considered for further examination by clinicians and scientists alike when designing and evaluating rocker profile footwear.

Future Studies

Through this study, we were able to answer some lingering questions regarding the rollover performance provided by rocker profile footwear. However, there are more questions that need to be addressed based on the results of this study. Most importantly, this study should be repeated with a greater sample size in order to provide more definitive results to quantify and characterize effective and ineffective rollover provided by footwear. Also based on the results of subjects' heart rate and rating of perceived exertion, the possible lower limb movements adapted during the STOP-NR condition need to be examined by analyzing mechanical work, joint moments and joint power.

By analyzing the ground reaction forces in tandem with the other rollover outputs (i.e., stance phase duration, cadence), we have shown that a cushioned heel rocker is necessary to produce effective rollover, however, this study has not determined the optimal stiffness of the cushioned heel necessary to produce effective rollover. A future study, using a modular version of the current design, which is focused on optimizing the stiffness of the heel portion of the rocker profile footwear may provide additional insights for shoe designers, clinicians and scientists. Finally, the footwear attached to the ankle-foot orthosis used in this study was designed to influence motions in the heel and midfoot (and this influence the heel rocker and ankle rocker functions of the foot and ankle complex). The forefoot was designed to minimally inhibit the forefoot rocker function by providing a forefoot fulcrum, fenestration and toe ramp that would retain propulsion by the subjects in late stance. As a result, the complete efficacy of the rocker profile in providing passive rollover while the ankle foot complex is constrained was not fully determined. A future study aimed at determining the optimal design and material properties for a completely passive heel to toe rocker profile system which provides effective rollover is a topic for future study. Persons without the ability to generate normal propulsive forces may be indicated for a fully passive heel to toe rocker profile system. Such a study will be possible based on the results of this study which is one of the first studies to characterize the outputs of that define effective rollover in a rocker profile footwear system.

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